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# ELECTROCHEMICAL ANALYTE SENSOR

## Field of the Invention

The present invention is, in general, directed to an analyte sensor. More particularly, the present invention relates to an electrochemical sensor for determining a level of an analyte, such as glucose, lactate, or oxygen, *in vivo* and/or *in vitro*.

## Background of the Invention

The monitoring of the level of glucose or other analytes, such as lactate or oxygen, in certain individuals is vitally important to their health. High or low levels of glucose or other analytes may have detrimental effects. The monitoring of glucose is particularly important to individuals with diabetes, as they must determine when insulin is needed to reduce glucose levels in their bodies or when additional glucose is needed to raise the level of glucose in their bodies.

A conventional technique used by many diabetics for personally monitoring their blood glucose level includes the periodic drawing of blood, the application of that blood to a test strip, and the determination of the blood glucose level using colorimetric, electrochemical, or photometric detection. This technique does not permit continuous or automatic monitoring of glucose levels in the body, but typically must be performed manually on a periodic basis. Unfortunately, the consistency with which the level of glucose is checked varies widely among individuals. Many diabetics find the periodic testing inconvenient and they sometimes forget to test their glucose level or do not have time for a proper test. In addition, some individuals may wish to avoid the pain associated with the test. These situations may result in hyperglycemic or hypoglycemic episodes. An *in vivo* glucose sensor that continuously or automatically monitors the

individual's glucose level would enable individuals to more easily monitor their glucose, or other analyte, levels.

A variety of devices have been developed for continuous or automatic monitoring of analytes, such as glucose, in the blood stream or interstitial fluid. Many of these devices use electrochemical sensors which are directly implanted into a blood vessel or in the subcutaneous tissue of a patient. However, these devices are often difficult to reproducibly and inexpensively manufacture in large numbers. In addition, these devices are typically large, bulky, and/or inflexible, and many can not be used effectively outside of a controlled medical facility, such as a hospital or a doctor's office, unless the patient is restricted in his activities.

The patient's comfort and the range of activities that can be performed while the sensor is implanted are important considerations in designing extended-use sensors for continuous or automatic *in vivo* monitoring of the level of an analyte, such as glucose. There is a need for a small, comfortable device which can continuously monitor the level of an analyte, such as glucose, while still permitting the patient to engage in normal activities. Continuous and/or automatic monitoring of the analyte can provide a warning to the patient when the level of the analyte is at or near a threshold level. For example, if glucose is the analyte, then the monitoring device might be configured to warn the patient of current or impending hyperglycemia or hypoglycemia. The patient can then take appropriate actions.

In addition to *in vivo* monitoring of analyte levels, it is often important to determine the level of an analyte in a sample taken from a subject. For many individuals and for many analytes, continuous monitoring of analyte level is not necessary, convenient, and/or desirable. *In vitro* measurements are often useful in making periodic determinations of analyte level when an *in vivo* sensor is not being used. Such measurements may also be useful for calibrating an *in vivo* sensor. In these cases, it may be desirable to use small volume samples due to the difficulty of obtaining such samples, the discomfort of the patient when the sample is obtained, and/or other reasons. However, most conventional sensors are designed to test for analyte levels in

samples larger than 3 microliters. It is desirable to have sensors that could be used for the *in vitro* monitoring of samples that may be as small as a microliter, or even 25 nanoliters or less. The use of such small samples reduces the inconvenience and pain associated with obtaining a sample, for example, by lancing a portion of the body to obtain a blood sample.

## Summary of the Invention

Generally, the present invention relates to an analyte sensor which can be used for the *in vivo* and/or *in vitro* determination of a level of an analyte in a fluid. Some embodiments of the invention are particularly useful for the continuous or automatic monitoring of a level of an analyte, such as glucose or lactate, in a patient. One embodiment of the invention is an electrochemical sensor. The electrochemical sensor includes a substrate, a recessed channel formed in a surface of the substrate, and a conductive material disposed in the recessed channel. The conductive material forms a working electrode.

Another embodiment of the invention is an electrochemical sensor that includes a substrate and a plurality of recessed channels formed in at least one surface of the substrate. Conductive material is disposed in each of the recessed channels. The conductive material in at least one of the recessed channels forms a working electrode.

A further embodiment of the invention is an analyte responsive electrochemical sensor that includes a working electrode and a mass transport limiting membrane. The mass transport limiting membrane preferably maintains a rate of permeation of the analyte through the mass transport limiting membrane with a variation of less than 3% per °C at temperatures ranging from 30°C to 40°C.

Yet another embodiment of the invention is a method of determining a level of an analyte in a fluid. The fluid is contacted by an electrochemical sensor that includes a substrate, a recessed channel in the substrate, and conductive material in the recessed channel forming a working electrode. An electrical signal is generated by the sensor in



4

response to the presence of the analyte. The level of the analyte may be determined from the electrical signal.

A further embodiment of the invention is a temperature sensor. The temperature sensor includes a substrate, a recessed channel formed in the substrate, and a temperature probe disposed in the recessed channel. The temperature probe includes two probe leads that are disposed in spaced-apart portions of the recessed channel and a temperature-dependent element that is disposed in the recessed channel and is in contact with the two probe leads. The temperature-dependent element is formed using a material having a temperature-dependent characteristic that alters a signal from the temperature probe in response to a change in temperature.

One embodiment of the invention is a method of determining a level of an analyte in a fluid. The fluid is placed in contact with an electrochemical sensor. The electrochemical sensor has a substrate, a recessed channel formed in a surface of the substrate, conductive material disposed in the recessed channel to form a working electrode, and a catalyst proximally disposed to the working electrode. A level of a second compound in the fluid is changed by a reaction of the analyte catalyzed by the catalyst. A signal is generated in response to the level of the second electrode. The level of the analyte is determined from the signal.

Another embodiment of the invention is an electrochemical sensor having a substrate and a working electrode disposed on the substrate. The working electrode preferably includes a carbon material and has a width along at least a portion of the working electrode of 150 μm or less.

Another embodiment of the invention is an electrochemical sensor for determining a level of an analyte in a fluid. The electrochemical sensor includes a substrate, a recessed channel formed in a surface of the substrate, and conductive material disposed in the recessed channel to form a working electrode. A catalyst is positioned near the working electrode to catalyze a reaction of the analyte which results in a change in a level of a second compound. The electrochemical sensor produces a signal which is responsive to the level of the second compound.

Yet another embodiment of the invention is a sensor adapted for subcutaneous implantation. The sensor includes a substrate, and conductive carbon non-leachably disposed on the substrate to form a working electrode. An enzyme is non-leachably disposed in proximity to the working electrode.

Another embodiment of the invention is an electrochemical sensor including a substrate and conductive material disposed on the substrate. The conductive material forms a plurality of traces. At least one of the traces forms a working electrode. The plurality of conductive traces are preferably separated on the surface of the substrate by a distance of 150 μm or less.

One embodiment of the invention is an electrochemical sensor having a substrate and conductive material disposed on a surface of the substrate. The conductive material forms a plurality of conductive traces, at least one of which forms a working electrode. The plurality of conductive traces are disposed on the surface of the substrate at a preferred density, along a width of the substrate, of 667 μm/trace or less.

Another embodiment of the invention is an electrochemical sensor having a substrate, a conductive material disposed on the substrate to form a working electrode, and a contact pad disposed on the substrate and operatively connected to the working electrode. The contact pad is made of a non-metallic conductive material to avoid or reduce corrosion.

Yet another embodiment of the invention is an analyte monitoring system having a sensor and a control unit. The sensor includes a substrate, a working electrode disposed on the substrate, and a contact pad coupled to the working electrode. The control unit has a conductive contact coupled to the working electrode and is configured to apply a potential across the working electrode. At least one of the contact pad and the conductive contact is made using a non-metallic material to avoid or reduce corrosion.

A further embodiment of the invention is a method of determining a level of an analyte in an animal. A sensor is implanted in the animal. The sensor includes a substrate, a plurality of conductive traces disposed on the substrate, and a working



6

electrode formed from one of the conductive traces. A signal is generated at the working electrode in response to the analyte. The level of the analyte is determined by analyzing the signal. If the level of the analyte exceeds a threshold amount, an electrical current is produced through a portion of the animal to warn the animal. The electrical current is produced by applying a potential between two of the conductive traces.

Another embodiment is an electrochemical sensor having a substrate, a conductive material disposed on the substrate to form a working electrode, and catalyst disposed in the conductive material. The catalyst catalyzes a reaction of the analyte to generate a signal at the working electrode.

The above summary of the present invention is not intended to describe each disclosed embodiment or every implementation of the present invention. The Figures and the detailed description which follow more particularly exemplify these embodiments.



## Brief Description of the Drawings

The invention may be more completely understood in consideration of the following detailed description of various embodiments of the invention in connection with the accompanying drawings, in which:

Figure 1 is a block diagram of one embodiment of an analyte monitor using an analyte sensor, according to the invention;

Figure 2 is a top view of one embodiment of an analyte sensor, according to the invention;

Figure 3A is a cross-sectional view of the analyte sensor of Figure 2;

Figure 3B is a cross-sectional view of another embodiment of an analyte sensor, according to the invention;

Figure 4A is a cross-sectional view of yet another embodiment of an analyte sensor, according to the invention;

Figure 4B is a cross-sectional view of a fourth embodiment of an analyte sensor, according to the invention;

Figure 5 is an expanded top view of a tip portion of the analyte sensor of Figure 2;

Figure 6 is a cross-sectional view of a fifth embodiment of an analyte sensor, according to the invention;

Figure 7 is an expanded top view of a tip-portion of the analyte sensor of Figure 6;

Figure 8 is an expanded bottom view of a tip-portion of the analyte sensor of Figure 6;

Figure 9 is a side view of the analyte sensor of Figure 2;

Figure 10 is a top view of the analyte sensor of Figure 6;

Figure 11 is a bottom view of the analyte sensor of Figure 6; and

Figure 12 is another embodiment of an analyte sensor according to the invention.

While the invention is amenable to various modifications and alternative forms, specifics thereof have been shown by way of example in the drawings and will be described in detail. It should be understood, however, that the intention is not to limit the invention to the particular embodiments described. On the contrary, the intention is to cover all modifications, equivalents, and alternatives falling within the spirit and scope of the invention.

## Detailed Description of the Preferred Embodiment

The present invention is applicable to an analyte sensor for the *in vivo* and/or *in vitro* determination of an analyte, such as glucose, lactate, or oxygen, in a fluid. The analyte sensors of the present invention can be utilized in a variety of contexts. For example, one embodiment of the analyte sensor can be subcutaneously implanted in the interstitial tissue of a patient for the continuous or periodic monitoring of a level of an analyte in a patient's interstitial fluid. This can then be used to infer the analyte level in the patient's bloodstream. Other *in vivo* analyte sensors can be made, according to the invention, for insertion into an organ, vein, artery, or other portion of the body

containing fluid. The *in vivo* analyte sensors may be configured for obtaining a single measurement and/or for monitoring the level of the analyte over a time period which may range from hours to days or longer.

Another embodiment of the analyte sensor can be used for the *in vitro* determination of the presence and/or level of an analyte in a sample, and, particularly, in a small volume sample (e.g., 10 microliters to 50 nanoliters or less). While the present invention is not so limited, an appreciation of various aspects of the invention may be gained through a discussion of the examples provided below.

The following definitions are provided for terms used herein. A "counter electrode" refers to an electrode paired with the working electrode, through which passes a current equal in magnitude and opposite in sign to the current passing through the working electrode. In the context of the invention, the term "counter electrode" is meant to include counter electrodes which also function as reference electrodes (i.e., a counter/reference electrode).

An "electrochemical sensor" is a device configured to detect the presence and/or measure the level of an analyte in a sample via electrochemical oxidation and reduction reactions on the sensor. These reactions are transduced to an electrical signal that can be correlated to an amount, concentration, or level of an analyte in the sample.

"Electrolysis" is the electrooxidation or electroreduction of a compound either directly at an electrode or via one or more electron transfer agents.

A compound is "immobilized" on a surface when it is entrapped on or chemically bound to the surface.

A "non-leachable" or "non-releasable" compound or a compound that is "non-leachably disposed" is meant to define a compound that is affixed on the sensor such that it does not substantially diffuse away from the working surface of the working electrode for the period in which the sensor is used (e.g., the period in which the sensor is implanted in a patient or measuring a sample).

Components are "immobilized" within a sensor, for example, when the components are covalently, ionically, or coordinatively bound to constituents of the

sensor and/or are entrapped in a polymeric or sol-gel matrix or membrane which precludes mobility.

An "electron transfer agent" is a compound that carries electrons between the analyte and the working electrode, either directly, or in cooperation with other electron transfer agents. One example of an electron transfer agent is a redox mediator.

A "working electrode" is an electrode at which the analyte (or a second compound whose level depends on the level of the analyte) is electrooxidized or electroreduced with or without the agency of an electron transfer agent.

A "working surface" is that portion of the working electrode which is coated with or is accessible to the electron transfer agent and configured for exposure to an analyte-containing fluid.

A "sensing layer" is a component of the sensor which includes constituents that facilitate the electrolysis of the analyte. The sensing layer may include constituents such as an electron transfer agent, a catalyst which catalyzes a reaction of the analyte to produce a response at the electrode, or both. In some embodiments of the sensor, the sensing layer is non-leachably disposed in proximity to or on the working electrode.

A "non-corroding" conductive material includes non-metallic materials, such as carbon and conductive polymers.

**Analyte Sensor Systems**

The sensors of the present invention can be utilized in a variety of devices and under a variety of conditions. The particular configuration of a sensor may depend on the use for which the sensor is intended and the conditions under which the sensor will operate (e.g., *in vivo* or *in vitro*). One embodiment of the analyte sensor is configured for implantation into a patient or user for *in vivo* operation. For example, implantation of the sensor may be made in the arterial or venous systems for direct testing of analyte levels in blood. Alternatively, a sensor may be implanted in the interstitial tissue for determining the analyte level in interstitial fluid. This level may be correlated and/or converted to analyte levels in blood or other fluids. The site and depth of implantation

may affect the particular shape, components, and configuration of the sensor. Subcutaneous implantation may be preferred, in some cases, to limit the depth of implantation of the sensor. Sensors may also be implanted in other regions of the body to determine analyte levels in other fluids.

An implantable analyte sensor may be used as part of an analyte monitoring system to continuously and/or periodically monitor the level of an analyte in a body fluid of a patient. In addition to the sensor 42, the analyte monitoring system 40 also typically includes a control unit 44 for operating the sensor 42 (e.g., providing a potential to the electrodes and obtaining measurements from the electrodes) and a processing unit 45 for analyzing the measurements from the sensor 42. The control unit 44 and processing unit 45 may be combined in a single unit or may be separate.

Sub C2 Another embodiment of the sensor may be used for *in vitro* measurement of a level of an analyte. The *in vitro* sensor is coupled to a control unit and/or a processing unit to form an analyte monitoring system. In some embodiments, an *in vitro* analyte monitoring system is also configured to provide a sample to the sensor. For example, the analyte monitoring system may be configured to draw a sample from, for example, a lanced wound using a wicking and/or capillary action. The sample may then be drawn into contact with the sensor. Examples of such sensors may be found in U.S. Patent Application Serial No. 08/795,767 and PCT Patent Application No. __________, entitled "Small Volume *in vitro* Analyte Sensor", filed on February 6, 1998, Attorney Docket No. 12008.11WO01, incorporated herein by reference.

Other methods for providing a sample to the sensor include using a pump, syringe, or other mechanism to draw a sample from a patient through tubing or the like either directly to the sensor or into a storage unit from which a sample is obtained for the sensor. The pump, syringe, or other mechanism may operate continuously, periodically, or when desired to obtain a sample for testing. Other useful devices for providing an analyte-containing fluid to the sensor include microfiltration and/or microdialysis devices. In some embodiments, particularly those using a microdialysis device, the analyte may be drawn from the body fluid through a microporous

membrane, for example, by osmotic pressure, into a carrier fluid which is then conveyed to the sensor for analysis. Other useful devices for acquiring a sample are those that collect body fluids transported across the skin using techniques, such as reverse iontophoresis, to enhance the transport of fluid containing analyte across the skin.

**The Sensor**

A sensor 42, according to the invention, includes at least one working electrode 58 formed on a substrate 50, as shown in Figure 2. The sensor 42 may also include at least one counter electrode 60 (or counter/reference electrode) and/or at least one reference electrode 62 (see Figure 8). The counter electrode 60 and/or reference electrode 62 may be formed on the substrate 50 or may be separate units. For example, the counter electrode and/or reference electrode may be formed on a second substrate which is also implanted in the patient or, for some embodiments of the implantable sensors, the counter electrode and/or reference electrode may be placed on the skin of the patient with the working electrode or electrodes being implanted into the patient. The use of an on-the-skin counter and/or reference electrode with an implantable working electrode is described in U.S. Patent No. 5,593, 852, incorporated herein by reference.

The working electrode or electrodes 58 are formed using conductive traces 52 disposed on the substrate 50. The counter electrode 60 and/or reference electrode 62, as well as other optional portions of the sensor 42, such as a temperature probe 66 (see Figure 8), may also be formed using conductive traces 52 disposed on the substrate 50. These conductive traces 52 may be formed over a smooth surface of the substrate 50 or within channels 54 formed by, for example, embossing, indenting or otherwise creating a depression in the substrate 50.

A sensing layer 64 (see Figures 3A and 3B) is often formed proximate to or on at least one of the working electrodes 58 to facilitate the electrochemical detection of the analyte and the determination of its level in the sample fluid, particularly if the analyte can not be electrolyzed at a desired rate and/or with a desired specificity on a



12

bare electrode. The sensing layer 64 may include an electron transfer agent to transfer electrons directly or indirectly between the analyte and the working electrode 58. The sensing layer 64 may also contain a catalyst to catalyze a reaction of the analyte. The components of the sensing layer may be in a fluid or gel that is proximate to or in contact with the working electrode 58. Alternatively, the components of the sensing layer 64 may be disposed in a polymeric or sol-gel matrix that is proximate to or on the working electrode 58. Preferably, the components of the sensing layer 64 are non-leachably disposed within the sensor 42. More preferably, the components of the sensor 42 are immobilized within the sensor 42.

In addition to the electrodes 58, 60, 62 and the sensing layer 64, the sensor 42 may also include a temperature probe 66 (see Figures 6 and 8), a mass transport limiting layer 74 (see Figure 9), a biocompatible layer 75 (see Figure 9), and/or other optional components, as described below. Each of these items enhances the functioning of and/or results from the sensor 42, as discussed below.



**The Substrate**

The substrate 50 may be formed using a variety of non-conducting materials, including, for example, polymeric or plastic materials and ceramic materials. Suitable materials for a particular sensor 42 may be determined, at least in part, based on the desired use of the sensor 42 and properties of the materials.

In some embodiments, the substrate is flexible. For example, if the sensor 42 is configured for implantation into a patient, then the sensor 42 may be made flexible (although rigid sensors may also be used for implantable sensors) to reduce pain to the patient and damage to the tissue caused by the implantation of and/or the wearing of the sensor 42. A flexible substrate 50 often increases the patient's comfort and allows a wider range of activities. A flexible substrate 50 is also useful for an *in vitro* sensor 42, particularly for ease of manufacturing. Suitable materials for a flexible substrate 50 include, for example, non-conducting plastic or polymeric materials and other non-conducting, flexible, deformable materials. Examples of useful plastic or polymeric

materials include thermoplastics such as polycarbonates, polyesters (e.g., Mylar™ and polyethylene terephthalate (PET)), polyvinyl chloride (PVC), polyurethanes, polyethers, polyamides, polyimides, or copolymers of these thermoplastics, such as PETG (glycol-modified polyethylene terephthalate).

In other embodiments, the sensors 42 are made using a relatively rigid substrate 50 to, for example, provide structural support against bending or breaking. Examples of rigid materials that may be used as the substrate 50 include poorly conducting ceramics, such as aluminum oxide and silicon dioxide. One advantage of an implantable sensor 42 having a rigid substrate is that the sensor 42 may have a sharp point and/or a sharp edge to aid in implantation of a sensor 42 without an additional insertion device. In addition, rigid substrates 50 may also be used in sensors for *in vitro* analyte monitors.

It will be appreciated that for many sensors 42 and sensor applications, both rigid and flexible sensors will operate adequately. The flexibility of the sensor 42 may also be controlled and varied along a continuum by changing, for example, the composition and/or thickness of the substrate 50.

In addition to considerations regarding flexibility, it is often desirable that implantable sensors 42, as well as *in vitro* sensors which contact a fluid that is returned to a patient's body, should have a substrate 50 which is non-toxic. Preferably, the substrate 50 is approved by one or more appropriate governmental agencies or private groups for *in vivo* use.

The sensor 42 may include optional features to facilitate insertion of an implantable sensor 42, as shown in Figure 12. For example, the sensor 42 may be pointed at the tip 123 to ease insertion. In addition, the sensor 42 may include a barb 125 which assists in anchoring the sensor 42 within the tissue of the patient during operation of the sensor 42. However, the barb 125 is typically small enough that little damage is caused to the subcutaneous tissue when the sensor 42 is removed for replacement.

Although the substrate 50 in at least some embodiments has uniform dimensions along the entire length of the sensor 42, in other embodiments, the substrate 50 has a



14

distal end 67 and a proximal end 65 with different widths 53, 55, respectively, as illustrated in Figure 2. In these embodiments, the distal end 67 of the substrate 50 may have a relatively narrow width 53. For sensors 42 which are implantable into the subcutaneous tissue or another portion of a patient's body, the narrow width 53 of the distal end 67 of the substrate 50 may facilitate the implantation of the sensor 42. Often, the narrower the width of the sensor 42, the less pain the patient will feel during implantation of the sensor and afterwards.

For subcutaneously implantable sensors 42 which are designed for continuous or periodic monitoring of the analyte during normal activities of the patient, a distal end 67 of the sensor 42 which is to be implanted into the patient has a width 53 of 2mm or less, preferably 1mm or less, and more preferably 0.5mm or less. If the sensor 42 does not have regions of different widths, then the sensor 42 will typically have an overall width of, for example, 2 mm, 1.5 mm, 1 mm, 0.5 mm, 0.25 mm, or less. However, wider or narrower sensors may be used. In particular, wider implantable sensors may be used for insertion into veins or arteries or when the movement of the patient is limited, for example, when the patient is confined in bed or in a hospital.

For sensors 42 which are designed for measuring small volume *in vitro* samples, the narrow width 53 may reduce the volume of sample needed for an accurate reading. The narrow width 53 of the sensor 42 results in all of the electrodes of the sensor 42 being closely congregated, thereby requiring less sample volume to cover all of the electrodes. The width of an *in vitro* sensor 42 may vary depending, at least in part, on the volume of sample available to the sensor 42 and the dimensions of the sample chamber in which the sensor 42 is disposed.

Returning to Figure 2, the proximal end 65 of the sensor 42 may have a width 55 larger than the distal end 67 to facilitate the connection between contact pads 49 of the electrodes and contacts on a control unit. The wider the sensor 42 at this point, the larger the contact pads 49 can be made. This may reduce the precision needed to properly connect the sensor 42 to contacts on the control unit (e.g., sensor control unit 44 of Figure 1). However, the maximum width of the sensor 42 may be constrained so

that the sensor 42 remains small for the convenience and comfort of the patient and/or to fit the desired size of the analyte monitor. For example, the proximal end 65 of a subcutaneously implantable sensor 42, such as the sensor 42 illustrated in Figure 1, may have a width 55 ranging from 0.5mm to 15mm, preferably from 1mm to 10mm, and more preferably from 3mm to 7mm. However, wider or narrower sensors may be used in this and other *in vivo* and *in vitro* applications.

The thickness of the substrate 50 may be determined by the mechanical properties of the substrate material (e.g., the strength, modulus, and/or flexibility of the material), the desired use of the sensor 42 including stresses on the substrate 50 arising from that use, as well as the depth of any channels or indentations formed in the substrate 50, as discussed below. Typically, the substrate 50 of a subcutaneously implantable sensor 42 for continuous or periodic monitoring of the level of an analyte while the patient engages in normal activities has a thickness of 50 to 500 μm and preferably 100 to 300 μm. However, thicker and thinner substrates 50 may be used, particularly in other types of *in vivo* and *in vitro* sensors 42.

The length of the sensor 42 may have a wide range of values depending on a variety of factors. Factors which influence the length of an implantable sensor 42 may include the depth of implantation into the patient and the ability of the patient to manipulate a small flexible sensor 42 and make connections between the sensor 42 and the sensor control unit 44. A subcutaneously implantable sensor 42 for the analyte monitor illustrated in Figure 1 may have a length ranging from 0.3 to 5 cm, however, longer or shorter sensors may be used. The length of the narrow portion of the sensor 42 (e.g., the portion which is subcutaneously inserted into the patient), if the sensor 42 has narrow and wide portions, is typically about 0.25 to 2 cm in length. However, longer and shorter portions may be used. All or only a part of this narrow portion may be subcutaneously implanted into the patient.

The lengths of other implantable sensors 42 will vary depending, at least in part, on the portion of the patient into which the sensor 42 is to be implanted or inserted. The length of *in vitro* sensors may vary over a wide range depending on the particular

configuration of the analyte monitoring system and, in particular, the distance between the contacts of the control unit and the sample.

**Conductive Traces**

At least one conductive trace 52 is formed on the substrate for use in constructing a working electrode 58. In addition, other conductive traces 52 may be formed on the substrate 50 for use as electrodes (e.g., additional working electrodes, as well as counter, counter/reference, and/or reference electrodes) and other components, such as a temperature probe. The conductive traces 52 may extend most of the distance along a length 57 of the sensor 50, as illustrated in Figure 2, although this is not necessary. The placement of the conductive traces 52 may depend on the particular configuration of the analyte monitoring system (e.g., the placement of control unit contacts and/or the sample chamber in relation to the sensor 42). For implantable sensors, particularly subcutaneously implantable sensors, the conductive traces typically extend close to the tip of the sensor 42 to minimize the amount of the sensor that must be implanted.

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The conductive traces 52 may be formed on the substrate 50 by a variety of techniques, including, for example, photolithography, screen printing, or other impact or non-impact printing techniques. The conductive traces 52 may also be formed by carbonizing conductive traces 52 in an organic (e.g., polymeric or plastic) substrate 50 using a laser. A description of some exemplary methods for forming the sensor 42 is provided in U.S. Patent Application Serial No. __________, entitled "Process for the Manufacture of an Electrochemical Biosensor", filed on March 4, 1998, Attorney Docket No. M&G 12008.16US01, incorporated herein by reference.

Another method for disposing the conductive traces 52 on the substrate 50 includes the formation of recessed channels 54 in one or more surfaces of the substrate 50 and the subsequent filling of these recessed channels 54 with a conductive material 56, as shown in Figure 3A. The recessed channels 54 may be formed by indenting, embossing, or otherwise creating a depression in the surface of the substrate 50.

Exemplary methods for forming channels and electrodes in a surface of a substrate can be found in U.S. Patent Application Serial No. ____________, entitled "Process for the Manufacture of an Electrochemical Biosensor", filed on March 4, 1998, Attorney Docket No. 12008.16US01. The depth of the channels is typically related to the thickness of the substrate 50. In one embodiment, the channels have depths in the range of about 12.5 to 75 μm ( 0.5 to 3 mils), and preferably about 25 to 50 μm (1 to 2 mils).

The conductive traces are typically formed using a conductive material 56 such as carbon (e.g., graphite), a conductive polymer, a metal or alloy (e.g., gold or gold alloy), or a metallic compound (e.g., ruthenium dioxide or titanium dioxide). The formation of films of carbon, conductive polymer, metal, alloy, or metallic compound are well-known and include, for example, chemical vapor deposition (CVD), physical vapor deposition, sputtering, reactive sputtering, printing, coating, and painting. The conductive material 56 which fills the channels 54 is often formed using a precursor material, such as a conductive ink or paste. In these embodiments, the conductive material 56 is deposited on the substrate 50 using methods such as coating, painting, or applying the material using a spreading instrument, such as a coating blade. Excess conductive material between the channels 54 is then removed by, for example, running a blade along the substrate surface.

In one embodiment, the conductive material 56 is a part of a precursor material, such as a conductive ink, obtainable, for example, from Ercon, Inc. (Wareham, MA), Metech, Inc. (Elverson, PA), E.I. du Pont de Nemours and Co. (Wilmington, DE), Emca-Remex Products (Montgomeryville, PA), or MCA Services (Melbourn, Great Britain). The conductive ink is typically applied as a semiliquid or paste which contains particles of the carbon, metal, alloy, or metallic compound and a solvent or dispersant. After application of the conductive ink on the substrate 50 (e.g., in the channels 54), the solvent or dispersant evaporates to leave behind a solid mass of conductive material 56.

In addition to the particles of carbon, metal, alloy, or metallic compound, the conductive ink may also contain a binder. The binder may optionally be cured to further bind the conductive material 56 within the channel 54 and/or on the substrate 50.

Curing the binder increases the conductivity of the conductive material 56. However, this is typically not necessary as the currents carried by the conductive material 56 within the conductive traces 52 are often relatively low (usually less than 1 μA and often less than 100 nA). Typical binders include, for example, polyurethane resins, cellulose derivatives, elastomers, and highly fluorinated polymers. Examples of elastomers include silicones, polymeric dienes, and acrylonitrile-butadiene-styrene (ABS) resins. One example of a fluorinated polymer binder is Teflon® (DuPont, Wilmington, DE). These binders are cured using, for example, heat or light, including ultraviolet (UV) light. The appropriate curing method typically depends on the particular binder which is used.

Often, when a liquid or semiliquid precursor of the conductive material 56 (e.g., a conductive ink) is deposited in the channel 54, the precursor fills the channel 54. However, when the solvent or dispersant evaporates, the conductive material 56 which remains may lose volume such that the conductive material 56 may or may not continue to fill the channel 54. Preferred conductive materials 56 do not pull away from the substrate 50 as they lose volume, but rather decrease in height within the channel 54. These conductive materials 56 typically adhere well to the substrate 50 and therefore do not pull away from the substrate 50 during evaporation of the solvent or dispersant. Other suitable conductive materials 56 either adhere to at least a portion of the substrate 50 and/or contain another additive, such as a binder, which adheres the conductive material 56 to the substrate 50. Preferably, the conductive material 56 in the channels 54 is non-leachable, and more preferably immobilized on the substrate 50. In some embodiments, the conductive material 56 may be formed by multiple applications of a liquid or semiliquid precursor interspersed with removal of the solvent or dispersant.

In another embodiment, the channels 54 are formed using a laser. The laser carbonizes the polymer or plastic material. The carbon formed in this process is used as the conductive material 56. Additional conductive material 56, such as a conductive carbon ink, may be used to supplement the carbon formed by the laser.

In a further embodiment, the conductive traces 52 are formed by pad printing techniques. For example, a film of conductive material is formed either as a continuous film or as a coating layer deposited on a carrier film. This film of conductive material is brought between a print head and the substrate 50. A pattern on the surface of the substrate 50 is made using the print head according to a desired pattern of conductive traces 52. The conductive material is transferred by pressure and/or heat from the film of conductive material to the substrate 50. This technique often produces channels (e.g., depressions caused by the print head) in the substrate 50. Alternatively, the conductive material is deposited on the surface of the substrate 50 without forming substantial depressions.

In other embodiments, the conductive traces 52 are formed by non-impact printing techniques. Such techniques include electrophotography and magnetography. In these processes, an image of the conductive traces 52 is electrically or magnetically formed on a drum. A laser or LED may be used to electrically form an image. A magnetic recording head may be used to magnetically form an image. A toner material (e.g., a conductive material, such as a conductive ink) is then attracted to portions of the drum according to the image. The toner material is then applied to the substrate by contact between the drum and the substrate. For example, the substrate may be rolled over the drum. The toner material may then be dried and/or a binder in the toner material may be cured to adhere the toner material to the substrate.

Another non-impact printing technique includes ejecting droplets of conductive material onto the substrate in a desired pattern. Examples of this technique include ink jet printing and piezo jet printing. An image is sent to the printer which then ejects the conductive material (e.g., a conductive ink) according to the pattern. The printer may provide a continuous stream of conductive material or the printer may eject the conductive material in discrete amounts at the desired points.

Yet another non-impact printing embodiment of forming the conductive traces includes an ionographic process. In the this process, a curable, liquid precursor, such as a photopolymerizable acrylic resin (e.g., Solimer 7501 from Cubital, Bad Kreuznach,





20

Germany) is deposited over a surface of a substrate 50. A photomask having a positive or negative image of the conductive traces 52 is then used to cure the liquid precursor. Light (e.g., visible or ultraviolet light) is directed through the photomask to cure the liquid precursor and form a solid layer over the substrate according to the image on the photomask. Uncured liquid precursor is removed leaving behind channels 54 in the solid layer. These channels 54 can then be filled with conductive material 56 to form conductive traces 52.

Conductive traces 52 (and channels 54, if used) can be formed with relatively narrow widths, for example, in the range of 25 to 250 μm, and including widths of, for example, 250μm, 150 μm, 100 μm, 75 μm, 50μm, 25μm or less by the methods described above. In embodiments with two or more conductive traces 52 on the same side of the substrate 50, the conductive traces 52 are separated by distances sufficient to prevent conduction between the conductive traces 52. The edge-to-edge distance between the conductive traces is preferably in the range of 25 to 250μm and may be, for example, 150 μm, 100 μm, 75μm, 50 μm, or less. The density of the conductive traces 52 on the substrate 50 is preferably in the range of about 150 to 700 μm/trace and may be as small as 667 μm/trace or less, 333 μm/trace or less, or even 167 μm/trace or less.

The working electrode 58 and the counter electrode 60 (if a separate reference electrode is used) are often made using a conductive material 56, such as carbon. Suitable carbon conductive inks are available from Ercon, Inc. (Wareham, MA), Metech, Inc. (Elverson, PA), E.I. du Pont de Nemours and Co. (Wilmington, DE), Emca-Remex Products (Montgomeryville, PA), or MCA Services (Melbourn, Great Britain). Typically, the working surface 51 of the working electrode 58 is at least a portion of the conductive trace 52 that is in contact with the analyte-containing fluid (e.g., implanted in the patient or in the sample chamber of an *in vitro* analyte monitor).

The reference electrode 62 and/or counter/reference electrode are typically formed using conductive material 56 that is a suitable reference material, for example silver/silver chloride or a non-leachable redox couple bound to a conductive material, for example, a carbon-bound redox couple. Suitable silver/silver chloride conductive

inks are available from Ercon, Inc. (Wareham, MA), Metech, Inc. (Elverson, PA), E.I. du Pont de Nemours and Co. (Wilmington, DE), Emca-Remex Products (Montgomeryville, PA), or MCA Services (Melbourn, Great Britain). Silver/silver chloride electrodes illustrate a type of reference electrode that involves the reaction of a metal electrode with a constituent of the sample or body fluid, in this case, $Cl^-$.

Suitable redox couples for binding to the conductive material of the reference electrode include, for example, redox polymers (e.g., polymers having multiple redox centers.) It is preferred that the reference electrode surface be non-corroding so that an erroneous potential is not measured. Preferred conductive materials include less corrosive metals, such as gold and palladium. Most preferred are non-corrosive materials including non-metallic conductors, such as carbon and conducting polymers. A redox polymer can be adsorbed on or covalently bound to the conductive material of the reference electrode, such as a carbon surface of a conductive trace 52. Non-polymeric redox couples can be similarly bound to carbon or gold surfaces.

A variety of methods may be used to immobilize a redox polymer on an electrode surface. One method is adsorptive immobilization. This method is particularly useful for redox polymers with relatively high molecular weights. The molecular weight of a polymer may be increased, for example, by cross-linking.

Another method for immobilizing the redox polymer includes the functionalization of the electrode surface and then the chemical bonding, often covalently, of the redox polymer to the functional groups on the electrode surface. One example of this type of immobilization begins with a poly(4-vinylpyridine). The polymer's pyridine rings are, in part, complexed with a reducible/oxidizable species, such as $[Os(bpy)_2Cl]^{+/2+}$ where bpy is 2,2'-bipyridine. Part of the pyridine rings are quaternized by reaction with 2-bromoethylamine. The polymer is then crosslinked, for example, using a diepoxide, such as polyethylene glycol diglycidyl ether.

Carbon surfaces can be modified for attachment of a redox species or polymer, for example, by electroreduction of a diazonium salt. As an illustration, reduction of a diazonium salt formed upon diazotization of p-aminobenzoic acid modifies a carbon

surface with phenylcarboxylic acid functional groups. These functional groups can then be activated by a carbodiimide, such as 1-ethyl-3-(3-dimethylaminopropyl)-carbodiimide hydrochloride. The activated functional groups are then bound with a amine-functionalized redox couple, such as the quaternized osmium-containing redox polymer described above or 2-aminoethylferrocene, to form the redox couple.

Similarly, gold can be functionalized by an amine, such as cystamine,. A redox couple such as $[Os(bpy)_2(pyridine\text{-}4\text{-}carboxylate)Cl]^{0/+}$ is activated by 1-ethyl-3-(3-dimethylaminopropyl)-carbodiimide hydrochloride to form a reactive O-acylisourea which reacts with the gold-bound amine to form an amide.

In one embodiment, in addition to using the conductive traces 52 as electrodes or probe leads, two or more of the conductive traces 52 on the substrate 50 are used to give the patient a mild electrical shock when, for example, the analyte level exceeds a threshold level. This shock may act as a warning or alarm to the patient to initiate some action to restore the appropriate level of the analyte.

The mild electrical shock is produced by applying a potential between any two conductive traces 52 that are not otherwise connected by a conductive path. For example, two of the electrodes 58, 60, 62 or one electrode 58, 60, 62 and the temperature probe 66 may be used to provide the mild shock. Preferably, the working electrode 58 and the reference electrode 62 are not used for this purpose as this may cause some damage to the chemical components on or proximate to the particular electrode (e.g., the sensing layer on the working electrode or the redox couple on the reference electrode).

The current used to produce the mild shock is typically 0.1 to 1 mA. Higher or lower currents may be used, although care should be taken to avoid harm to the patient. The potential between the conductive traces is typically 1 to 10 volts. However, higher or lower voltages may be used depending, for example, on the resistance of the conductive traces 52, the distance between the conductive traces 52 and the desired amount of current. When the mild shock is delivered, potentials at the working electrode 58 and across the temperature probe 66 may be removed to prevent harm to

those components caused by unwanted conduction between the working electrode 58 (and/or temperature probe 66, if used) and the conductive traces 52 which provide the mild shock.

**Contact Pads**

Typically, each of the conductive traces 52 includes a contact pad 49. The contact pad 49 may simply be a portion of the conductive trace 52 that is indistinguishable from the rest of the trace 52 except that the contact pad 49 is brought into contact with the conductive contacts of a control unit (e.g., the sensor control unit 44 of Figure 1). More commonly, however, the contact pad 49 is a region of the conductive trace 52 that has a larger width than other regions of the trace 52 to facilitate a connection with the contacts on the control unit. By making the contact pads 49 relatively large as compared with the width of the conductive traces 52, the need for precise registration between the contact pads 49 and the contacts on the control unit is less critical than with small contact pads.

The contact pads 49 are typically made using the same material as the conductive material 56 of the conductive traces 52. However, this is not necessary. Although metal, alloys, and metallic compounds may be used to form the contact pads 49, in some embodiments, it is desirable to make the contact pads 49 from a carbon or other non-metallic material, such as a conducting polymer. In contrast to metal or alloy contact pads, carbon and other non-metallic contact pads are not easily corroded if the contact pads 49 are in a wet, moist, or humid environment. Metals and alloys may corrode under these conditions, particularly if the contact pads 49 and contacts of the control unit are made using different metals or alloys. However, carbon and non-metallic contact pads 49 do not significantly corrode, even if the contacts of the control device are metal or alloy.

One embodiment of the invention includes a sensor 42 having contact pads 49 and a control unit 44 having conductive contacts (not shown). During operation of the sensor 42, the contact pads 49 and conductive contacts are in contact with each other.

In this embodiment, either the contact pads 49 or the conductive contacts are made using a non-corroding, conductive material. Such materials include, for example, carbon and conducting polymers. Preferred non-corroding materials include graphite and vitreous carbon. The opposing contact pad or conductive contact is made using carbon, a conducting polymer, a metal, such as gold, palladium, or platinum group metal, or a metallic compound, such as ruthenium dioxide. This configuration of contact pads and conductive contacts typically reduces corrosion. Preferably, when the sensor is placed in a 3 mM, and more preferably, in a 100 mM, NaCl solution, the signal arising due to the corrosion of the contact pads and/or conductive contacts is less than 3% of the signal generated by the sensor when exposed to concentration of analyte in the normal physiological range. For at least some subcutaneous glucose sensors, the current generated by analyte in a normal physiological range ranges from 3 to 500 nA.

Each of the electrodes 58, 60, 62, as well as the two probe leads 68, 70 of the temperature probe 66 (described below), are connected to contact pads 49 as shown in Figures 10 and 11. In one embodiment (not shown), the contact pads 49 are on the same side of the substrate 50 as the respective electrodes or temperature probe leads to which the contact pads 49 are attached.

In other embodiments, the conductive traces 52 on at least one side are connected through vias in the substrate to contact pads 49a on the opposite surface of the substrate 50, as shown in Figures 10 and 11. An advantage of this configuration is that contact between the contacts on the control unit and each of the electrodes 58, 60, 62 and the probe leads 68,70 of the temperature probe 66 can be made from a single side of the substrate 50.

In yet other embodiments (not shown), vias through the substrate are used to provide contact pads on both sides of the substrate 50 for each conductive trace 52. The vias connecting the conductive traces 52 with the contact pads 49a can be formed by making holes through the substrate 50 at the appropriate points and then filling the holes with conductive material 56.

**Exemplary Electrode Configurations**

A number of exemplary electrode configurations are described below, however, it will be understood that other configurations may also be used. In one embodiment, illustrated in Figure 3A, the sensor 42 includes two working electrodes 58a, 58b and one counter electrode 60, which also functions as a reference electrode. In another embodiment, the sensor includes one working electrode 58a, one counter electrode 60, and one reference electrode 62, as shown in Figure 3B. Each of these embodiments is illustrated with all of the electrodes formed on the same side of the substrate 50.

Alternatively, one or more of the electrodes may be formed on an opposing side of the substrate 50. This may be convenient if the electrodes are formed using two different types of conductive material 56 (e.g., carbon and silver/silver chloride). Then, at least in some embodiments, only one type of conductive material 56 needs to be applied to each side of the substrate 50, thereby reducing the number of steps in the manufacturing process and/or easing the registration constraints in the process. For example, if the working electrode 58 is formed using a carbon-based conductive material 56 and the reference or counter/reference electrode is formed using a silver/silver chloride conductive material 56, then the working electrode and reference or counter/reference electrode may be formed on opposing sides of the substrate 50 for ease of manufacture.

In another embodiment, two working electrodes 58 and one counter electrode 60 are formed on one side of the substrate 50 and one reference electrode 62 and a temperature probe 66 are formed on an opposing side of the substrate 50, as illustrated in Figure 6. The opposing sides of the tip of this embodiment of the sensor 42 are illustrated in Figures 7 and 8.

**Sensing Layer**

Some analytes, such as oxygen, can be directly electrooxidized or electroreduced on the working electrode 58. Other analytes, such as glucose and lactate, require the presence of at least one electron transfer agent and/or at least one catalyst to facilitate

the electrooxidation or electroreduction of the analyte. Catalysts may also be used for those analyte, such as oxygen, that can be directly electrooxidized or electroreduced on the working electrode 58. For these analytes, each working electrode 58 has a sensing layer 64 formed proximate to or on a working surface of the working electrode 58. Typically, the sensing layer 64 is formed near or on only a small portion of the working electrode 58, often near a tip of the sensor 42. This limits the amount of material needed to form the sensor 42 and places the sensing layer 64 in the best position for contact with the analyte-containing fluid (e.g., a body fluid, sample fluid, or carrier fluid).

The sensing layer 64 includes one or more components designed to facilitate the electrolysis of the analyte. The sensing layer 64 may include, for example, a catalyst to catalyze a reaction of the analyte and produce a response at the working electrode 58, an electron transfer agent to indirectly or directly transfer electrons between the analyte and the working electrode 58, or both.

The sensing layer 64 may be formed as a solid composition of the desired components (e.g., an electron transfer agent and/or a catalyst). These components are preferably non-leachable from the sensor 42 and more preferably are immobilized on the sensor 42. For example, the components may be immobilized on a working electrode 58. Alternatively, the components of the sensing layer 64 may be immobilized within or between one or more membranes or films disposed over the working electrode 58 or the components may be immobilized in a polymeric or sol-gel matrix. Examples of immobilized sensing layers are described in U.S. Patents Nos. 5,262,035, 5,264,104, 5,264,105, 5,320,725, 5,593,852, and 5,665,222, U.S. Patent Application No. 08/540,789, and PCT Patent Application No. _______ entitled "Soybean Peroxidase Electrochemical Sensor", filed on February 11, 1998, Attorney Docket No. M&G 12008.8WOI2, incorporated herein by reference.

In some embodiments, one or more of the components of the sensing layer 64 may be solvated, dispersed, or suspended in a fluid within the sensing layer 64, instead of forming a solid composition. The fluid may be provided with the sensor 42 or may

be absorbed by the sensor 42 from the analyte-containing fluid. Preferably, the components which are solvated, dispersed, or suspended in this type of sensing layer 64 are non-leachable from the sensing layer. Non-leachability may be accomplished, for example, by providing barriers(e.g., the electrode, substrate, membranes, and/or films) around the sensing layer which prevent the leaching of the components of the sensing layer 64. One example of such a barrier is a microporous membrane or film which allows diffusion of the analyte into the sensing layer 64 to make contact with the components of the sensing layer 64, but reduces or eliminates the diffusion of the sensing layer components (e.g., a electron transfer agent and/or a catalyst) out of the sensing layer 64.

A variety of different sensing layer configurations can be used. In one embodiment, the sensing layer 64 is deposited on the conductive material 56 of a working electrode 58a, as illustrated in Figures 3A and 3B. The sensing layer 64 may extend beyond the conductive material 56 of the working electrode 58a. In some cases, the sensing layer 64 may also extend over the counter electrode 60 or reference electrode 62 without degrading the performance of the glucose sensor. For those sensors 42 which utilize channels 54 within which the conductive material 56 is deposited, a portion of the sensing layer 64 may be formed within the channel 54 if the conductive material 56 does not fill the channel 54.

A sensing layer 64 in direct contact with the working electrode 58a may contain an electron transfer agent to transfer electrons directly or indirectly between the analyte and the working electrode, as well as a catalyst to facilitate a reaction of the analyte. For example, a glucose, lactate, or oxygen electrode may be formed having a sensing layer which contains a catalyst, such as glucose oxidase, lactate oxidase, or laccase, respectively, and an electron transfer agent that facilitates the electrooxidation of the glucose, lactate, or oxygen, respectively.

In another embodiment, the sensing layer 64 is not deposited directly on the working electrode 58a. Instead, the sensing layer 64 is spaced apart from the working electrode 58a, as illustrated in Figure 4A, and separated from the working electrode 58a

by a separation layer 61. The separation layer 61 typically includes one or more membranes or films. In addition to separating the working electrode 58a from the sensing layer 64, the separation layer 61 may also act as a mass transport limiting layer or an interferent eliminating layer, as described below.

Typically, a sensing layer 64, which is not in direct contact with the working electrode 58a, includes a catalyst that facilitates a reaction of the analyte. However, this sensing layer 64 typically does not include an electron transfer agent that transfers electrons directly from the working electrode 58a to the analyte, as the sensing layer 64 is spaced apart from the working electrode 58a. One example of this type of sensor is a glucose or lactate sensor which includes an enzyme (e.g., glucose oxidase or lactate oxidase, respectively) in the sensing layer 64. The glucose or lactate reacts with a second compound (e.g., oxygen) in the presence of the enzyme. The second compound is then electrooxidized or electroreduced at the electrode. Changes in the signal at the electrode indicate changes in the level of the second compound in the fluid and are proportional to changes in glucose or lactate level and, thus, correlate to the analyte level.

In another embodiment, two sensing layers 63, 64 are used, as shown in Figure 4B. Each of the two sensing layers 63, 64 may be independently formed on the working electrode 58a or in proximity to the working electrode 58a. One sensing layer 64 is typically, although not necessarily, spaced apart from the working electrode 58a. For example, this sensing layer 64 may include a catalyst which catalyzes a reaction of the analyte to form a product compound. The product compound is then electrolyzed in the second sensing layer 63 which may include an electron transfer agent to transfer electrons between the working electrode 58a and the product compound and/or a second catalyst to catalyze a reaction of the product compound to generate a signal at the working electrode 58a.

sub C6 For example, a glucose or lactate sensor may include a first sensing layer 64 which is spaced apart from the working electrode and contains an enzyme, for example, glucose oxidase or lactate oxidase. The reaction of glucose or lactate in the presence of



29

the appropriate enzyme forms hydrogen peroxide. A second sensing layer 63 is provided directly on the working electrode 58a and contains a peroxidase enzyme and an electron transfer agent to generate a signal at the electrode in response to the hydrogen peroxide. The level of hydrogen peroxide indicated by the sensor then correlates to the level of glucose or lactate. Another sensor which operates similarly can be made using a single sensing layer with both the glucose or lactate oxidase and the peroxidase being deposited in the single sensing layer. Examples of such sensors are described in U.S. Patent No. 5,593,852, U.S. Patent Application No. 08/540,789, and PCT Patent Application No. _______ entitled "Soybean Peroxidase Electrochemical Sensor", filed on February 11, 1998, Attorney Docket No. M&G 12008.8WOI2, incorporated herein by reference.

In some embodiments, one or more of the working electrodes 58b do not have a corresponding sensing layer 64, as shown in Figures 3A and 4A, or have a sensing layer (not shown) which does not contain one or more components (e.g., an electron transfer agent or catalyst) needed to electrolyze the analyte. The signal generated at this working electrode 58b typically arises from interferents and other sources, such as ions, in the fluid, and not in response to the analyte (because the analyte is not electrooxidized or electroreduced). Thus, the signal at this working electrode 58b corresponds to a background signal. The background signal can be removed from the analyte signal obtained from other working electrodes 58a that are associated with fully-functional sensing layers 64 by, for example, subtracting the signal at working electrode 58b from the signal at working electrode 58a.

Sensors having multiple working electrodes 58a may also be used to obtain more precise results by averaging the signals or measurements generated at these working electrodes 58a. In addition, multiple readings at a single working electrode 58a or at multiple working electrodes may be averaged to obtain more precise data.

**Electron Transfer Agent**

In many embodiments, the sensing layer 64 contains one or more electron transfer agents in contact with the conductive material 56 of the working electrode 58, as shown in Figures 3A and 3B. In some embodiments, it is acceptable for the electron transfer agent to diffuse or leach away from the working electrode, particularly for *in vitro* sensors 42 that are used only once. Other *in vitro* sensors may utilize a carrier fluid which contains the electron transfer agent. The analyte is transferred to the carrier fluid from the original sample fluid by, for example, osmotic flow through a microporous membrane or the like.

In yet other embodiments of the invention, there is little or no leaching of the electron transfer agent away from the working electrode 58 during the period in which the sensor 42 is implanted in the patient or measuring an *in vitro* analyte-containing sample. A diffusing or leachable (i.e., releasable) electron transfer agent often diffuses into the analyte-containing fluid, thereby reducing the effectiveness of the electrode by reducing the sensitivity of the sensor over time. In addition, a diffusing or leaching electron transfer agent in an implantable sensor 42 may also cause damage to the patient. In these embodiments, preferably, at least 90%, more preferably, at least 95%, and, most preferably, at least 99%, of the electron transfer agent remains disposed on the sensor after immersion in the analyte-containing fluid for 24 hours, and, more preferably, for 72 hours. In particular, for an implantable sensor, preferably, at least 90%, more preferably, at least 95%, and most preferably, at least 99%, of the electron transfer agent remains disposed on the sensor after immersion in the body fluid at 37°C for 24 hours, and, more preferably, for 72 hours.

In some embodiments of the invention, to prevent leaching, the electron transfer agents are bound or otherwise immobilized on the working electrode 58 or between or within one or more membranes or films disposed over the working electrode 58. The electron transfer agent may be immobilized on the working electrode 58 using, for example, a polymeric or sol-gel immobilization technique. Alternatively, the electron transfer agent may be chemically (e.g., ionically, covalently, or coordinatively) bound to

the working electrode 58, either directly or indirectly through another molecule, such as a polymer, that is in turn bound to the working electrode 58.

Application of the sensing layer 64 on a working electrode 58a is one method for creating a working surface for the working electrode 58a, as shown in Figures 3A and 3B. The electron transfer agent mediates the transfer of electrons to electrooxidize or electroreduce an analyte and thereby permits a current flow between the working electrode 58 and the counter electrode 60 via the analyte. The mediation of the electron transfer agent facilitates the electrochemical analysis of analytes which are not suited for direct electrochemical reaction on an electrode.

In general, the preferred electron transfer agents are electroreducible and electrooxidizable ions or molecules having redox potentials that are a few hundred millivolts above or below the redox potential of the standard calomel electrode (SCE). Preferably, the electron transfer agents are not more reducing than about -150 mV and not more oxidizing than about +400 mV versus SCE.

The electron transfer agent may be organic, organometallic, or inorganic. Examples of organic redox species are quinones and species that in their oxidized state have quinoid structures, such as Nile blue and indophenol. Some quinones and partially oxidized quinhydrones react with functional groups of proteins such as the thiol groups of cysteine, the amine groups of lysine and arginine, and the phenolic groups of tyrosine which may render those redox species unsuitable for some of the sensors of the present invention because of the presence of the interfering proteins in an analyte-containing fluid. Usually substituted quinones and molecules with quinoid structure are less reactive with proteins and are preferred. A preferred tetrasubstituted quinone usually has carbon atoms in positions 1, 2, 3, and 4.

In general, electron transfer agents suitable for use in the invention have structures or charges which prevent or substantially reduce the diffusional loss of the electron transfer agent during the period of time that the sample is being analyzed. The preferred electron transfer agents include a redox species bound to a polymer which can in turn be immobilized on the working electrode. The bond between the redox species



32

and the polymer may be covalent, coordinative, or ionic. Useful electron transfer agents and methods for producing them are described in U.S. Patent Nos. 5,264,104; 5,356,786; 5,262,035; and 5,320,725, incorporated herein by reference. Although any organic or organometallic redox species can be bound to a polymer and used as an electron transfer agent, the preferred redox species is a transition metal compound or complex. The preferred transition metal compounds or complexes include osmium, ruthenium, iron, and cobalt compounds or complexes. The most preferred are osmium compounds and complexes. It will be recognized that many of the redox species described below may also be used, typically without a polymeric component, as electron transfer agents in a carrier fluid or in a sensing layer of a sensor where leaching of the electron transfer agent is acceptable.

One type of non-releasable polymeric electron transfer agent contains a redox species covalently bound in a polymeric composition. An example of this type of mediator is poly(vinylferrocene).

Another type of non-releasable electron transfer agent contains an ionically-bound redox species. Typically, this type of mediator includes a charged polymer coupled to an oppositely charged redox species. Examples of this type of mediator include a negatively charged polymer such as Nafion® (DuPont) coupled to a positively charged redox species such as an osmium or ruthenium polypyridyl cation. Another example of an ionically-bound mediator is a positively charged polymer such as quaternized poly(4-vinyl pyridine) or poly(1-vinyl imidazole) coupled to a negatively charged redox species such as ferricyanide or ferrocyanide. The preferred ionically-bound redox species is a highly charged redox species bound within an oppositely charged redox polymer.

In another embodiment of the invention, suitable non-releasable electron transfer agents include a redox species coordinatively bound to a polymer. For example, the mediator may be formed by coordination of an osmium or cobalt 2, 2'-bipyridyl complex to poly(1-vinyl imidazole) or poly(4-vinyl pyridine).

The preferred electron transfer agents are osmium transition metal complexes with one or more ligands, each ligand having a nitrogen-containing heterocycle such as 2,2'-bipyridine, 1,10-phenanthroline, or derivatives thereof. Furthermore, the preferred electron transfer agents also have one or more ligands covalently bound in a polymer, each ligand having at least one nitrogen-containing heterocycle, such as pyridine, imidazole, or derivatives thereof. These preferred electron transfer agents exchange electrons rapidly between each other and the working electrodes 58 so that the complex can be rapidly oxidized and reduced.

One example of a particularly useful electron transfer agent includes (a) a polymer or copolymer having pyridine or imidazole functional groups and (b) osmium cations complexed with two ligands, each ligand containing 2,2'-bipyridine, 1,10-phenanthroline, or derivatives thereof, the two ligands not necessarily being the same. Preferred derivatives of 2,2'-bipyridine for complexation with the osmium cation are 4,4'-dimethyl-2,2'-bipyridine and mono-, di-, and polyalkoxy-2,2'-bipyridines, such as 4,4'-dimethoxy-2,2'-bipyridine. Preferred derivatives of 1,10-phenanthroline for complexation with the osmium cation are 4,7-dimethyl-1,10-phenanthroline and mono, di-, and polyalkoxy-1,10-phenanthrolines, such as 4,7-dimethoxy-1,10-phenanthroline. Preferred polymers for complexation with the osmium cation include polymers and copolymers of poly(1-vinyl imidazole) (referred to as "PVI") and poly(4-vinyl pyridine) (referred to as "PVP"). Suitable copolymer substituents of poly(1-vinyl imidazole) include acrylonitrile, acrylamide, and substituted or quaternized N-vinyl imidazole. Most preferred are electron transfer agents with osmium complexed to a polymer or copolymer of poly(1-vinyl imidazole).

The preferred electron transfer agents have a redox potential ranging from -100 mV to about +150 mV versus the standard calomel electrode (SCE). Preferably, the potential of the electron transfer agent ranges from -100 mV to +150 mV and more preferably, the potential ranges from -50 mV to +50 mV. The most preferred electron transfer agents have osmium redox centers and a redox potential ranging from +50 mV to -150 mV versus SCE.

**Catalyst**

The sensing layer 64 may also include a catalyst which is capable of catalyzing a reaction of the analyte. The catalyst may also, in some embodiments, act as an electron transfer agent. One example of a suitable catalyst is an enzyme which catalyzes a reaction of the analyte. For example, a catalyst, such as a glucose oxidase, glucose dehydrogenase (e.g., pyrroloquinoline quinone glucose dehydrogenase (PQQ)), or oligosaccharide dehydrogenase, may be used when the analyte is glucose. A lactate oxidase or lactate dehydrogenase may be used when the analyte is lactate. Laccase may be used when the analyte is oxygen or when oxygen is generated or consumed in response to a reaction of the analyte.

Preferably, the catalyst is non-leachably disposed on the sensor, whether the catalyst is part of a solid sensing layer in the sensor or solvated in a fluid within the sensing layer. More preferably, the catalyst is immobilized within the sensor (e.g., on the electrode and/or within or between a membrane or film) to prevent unwanted leaching of the catalyst away from the working electrode 58 and into the patient. This may be accomplished, for example, by attaching the catalyst to a polymer, cross linking the catalyst with another electron transfer agent (which, as described above, can be polymeric), and/or providing one or more barrier membranes or films with pore sizes smaller than the catalyst.

As described above, a second catalyst may also be used. This second catalyst is often used to catalyze a reaction of a product compound resulting from the catalyzed reaction of the analyte. The second catalyst typically operates with an electron transfer agent to electrolyze the product compound to generate a signal at the working electrode. Alternatively, the second catalyst may be provided in an interferent-eliminating layer to catalyze reactions that remove interferents, as described below.

One embodiment of the invention is an electrochemical sensor in which the catalyst is mixed or dispersed in the conductive material 56 which forms the conductive trace 52 of a working electrode 58. This may be accomplished, for example, by mixing

a catalyst, such as an enzyme, in a carbon ink and applying the mixture into a channel 54 on the surface of the substrate 50. Preferably, the catalyst is immobilized in the channel 53 so that it can not leach away from the working electrode 58. This may be accomplished, for example, by curing a binder in the carbon ink using a curing technique appropriate to the binder. Curing techniques include, for example, evaporation of a solvent or dispersant, exposure to ultraviolet light, or exposure to heat. Typically, the mixture is applied under conditions that do not substantially degrade the catalyst. For example, the catalyst may be an enzyme that is heat-sensitive. The enzyme and conductive material mixture should be applied and cured, preferably, without sustained periods of heating. The mixture may be cured using evaporation or UV curing techniques or by the exposure to heat that is sufficiently short that the catalyst is not substantially degraded.

Sub C17

Another consideration for *in vivo* analyte sensors is the thermostability of the catalyst. Many enzymes have only limited stability at biological temperatures. Thus, it may be necessary to use large amounts of the catalyst and/or use a catalyst that is thermostable at the necessary temperature (e.g., 37°C or higher for normal body temperature). A thermostable catalyst may be defined as a catalyst which loses less than 5% of its activity when held at 37°C for at least one hour, preferably, at least one day, and more preferably at least three days. One example of a thermostable catalyst is soybean peroxidase. This particular thermostable catalyst may be used in a glucose or lactate sensor when combined either in the same or separate sensing layers with glucose or lactate oxidase or dehydrogenase. A further description of thermostable catalysts and their use in electrochemical inventions is found in U.S. Patent No. 5,665,222 U.S. Patent Application No. 08/540,789, and PCT Application No. _______ entitle "Soybean Peroxidase Electrochemical Sensor", filed on February 11, 1998, Attorney Docket No. M&G 12008.8WOI2.

**Electrolysis of the Analyte**

To electrolyze the analyte, a potential (versus a reference potential) is applied across the working and counter electrodes 58, 60. The minimum magnitude of the applied potential is often dependent on the particular electron transfer agent, analyte (if the analyte is directly electrolyzed at the electrode), or second compound (if a second compound, such as oxygen or hydrogen peroxide, whose level is dependent on the analyte level, is directly electrolyzed at the electrode). The applied potential usually equals or is more oxidizing or reducing, depending on the desired electrochemical reaction, than the redox potential of the electron transfer agent, analyte, or second compound, whichever is directly electrolyzed at the electrode. The potential at the working electrode is typically large enough to drive the electrochemical reaction to or near completion.

The magnitude of the potential may optionally be limited to prevent significant (as determined by the current generated in response to the analyte) electrochemical reaction of interferents, such as urate, ascorbate, and acetaminophen. The limitation of the potential may be obviated if these interferents have been removed in another way, such as by providing an interferent-limiting barrier, as described below, or by including a working electrode 58b (see Figure 3A) from which a background signal may be obtained.

When a potential is applied between the working electrode 58 and the counter electrode 60, an electrical current will flow. The current is a result of the electrolysis of the analyte or a second compound whose level is affected by the analyte. In one embodiment, the electrochemical reaction occurs via an electron transfer agent and the optional catalyst. Many analytes B are oxidized (or reduced) to products C by an electron transfer agent species A in the presence of an appropriate catalyst (e.g., an enzyme). The electron transfer agent A is then oxidized (or reduced) at the electrode. Electrons are collected by (or removed from) the electrode and the resulting current is measured. This process is illustrated by reaction equations (1) and (2) (similar

equations may be written for the reduction of the analyte B by a redox mediator A in the presence of a catalyst):

T0380

$$nA(ox) + B \xrightarrow{\text{catalyst}} nA(red) + C \qquad (1)$$

$$nA(red) \xrightarrow{\text{electrode}} nA(ox) + ne^- \qquad (2)$$

As an example, an electrochemical sensor may be based on the reaction of a glucose molecule with two non-leachable ferricyanide anions in the presence of glucose oxidase to produce two non-leachable ferrocyanide anions, two hydrogen ions, and gluconolactone. The amount of glucose present is assayed by electrooxidizing the non-leachable ferrocyanide anions to non-leachable ferricyanide anions and measuring the current.

In another embodiment, a second compound whose level is affected by the analyte is electrolyzed at the working electrode. In some cases, the analyte D and the second compound, in this case, a reactant compound E, such as oxygen, react in the presence of the catalyst, as shown in reaction equation (3).

T0381

$$D + E \xrightarrow{\text{catalyst}} F + G \qquad (3)$$

The reactant compound E is then directly oxidized (or reduced) at the working electrode, as shown in reaction equation (4)

T0382

$$nE(red) \xrightarrow{\text{electrode}} nE(ox) + ne^- \qquad (4)$$

Alternatively, the reactant compound E is indirectly oxidized (or reduced) using an electron transfer agent H (optionally in the presence of a catalyst), that is subsequently reduced or oxidized at the electrode, as shown in reaction equations (5) and (6).

T0390

$$nH(ox) + E \longrightarrow nH(red) + I \qquad (5)$$

$$nH(red) \xrightarrow{\text{electrode}} nH(ox) + ne^- \qquad (6)$$

In either case, changes in the concentration of the reactant compound, as indicated by the signal at the working electrode, correspond inversely to changes in the analyte (i.e., as the level of analyte increase then the level of reactant compound and the signal at the electrode decreases.)

In other embodiments, the relevant second compound is a product compound F, as shown in reaction equation (3). The product compound F is formed by the catalyzed reaction of analyte D and then be directly electrolyzed at the electrode or indirectly electrolyzed using an electron transfer agent and, optionally, a catalyst. In these embodiments, the signal arising from the direct or indirect electrolysis of the product compound F at the working electrode corresponds directly to the level of the analyte (unless there are other sources of the product compound). As the level of analyte increases, the level of the product compound and signal at the working electrode increases.

Those skilled in the art will recognize that there are many different reactions that will achieve the same result; namely the electrolysis of an analyte or a compound whose level depends on the level of the analyte. Reaction equations (1) through (6) illustrate non-limiting examples of such reactions.

**Temperature Probe**

A variety of optional items may be included in the sensor. One optional item is a temperature probe 66 (Figures 8 and 11). The temperature probe 66 may be made using a variety of known designs and materials. One exemplary temperature probe 66 is formed using two probe leads 68, 70 connected to each other through a temperature-dependent element 72 that is formed using a material with a temperature-dependent

characteristic. An example of a suitable temperature-dependent characteristic is the resistance of the temperature-dependent element 72.

The two probe leads 68, 70 are typically formed using a metal, an alloy, a semimetal, such as graphite, a degenerate or highly doped semiconductor, or a small-band gap semiconductor. Examples of suitable materials include gold, silver, ruthenium oxide, titanium nitride, titanium dioxide, indium doped tin oxide, tin doped indium oxide, or graphite. The temperature-dependent element 72 is typically made using a fine trace (e.g., a conductive trace that has a smaller cross-section than that of the probe leads 68, 70) of the same conductive material as the probe leads, or another material such as a carbon ink, a carbon fiber, or platinum, which has a temperature-dependent characteristic, such as resistance, that provides a temperature-dependent signal when a voltage source is attached to the two probe leads 68, 70 of the temperature probe 66. The temperature-dependent characteristic of the temperature-dependent element 72 may either increase or decrease with temperature. Preferably, the temperature dependence of the characteristic of the temperature-dependent element 72 is approximately linear with temperature over the expected range of biological temperatures (about 25 to 45 °C), although this is not required.

Typically, a signal (e.g., a current) having an amplitude or other property that is a function of the temperature can be obtained by providing a potential across the two probe leads 68, 70 of the temperature probe 66. As the temperature changes, the temperature-dependent characteristic of the temperature-dependent element 72 increases or decreases with a corresponding change in the signal amplitude. The signal from the temperature probe 66 (e.g., the amount of current flowing through the probe) may be combined with the signal obtained from the working electrode 58 by, for example, scaling the temperature probe signal and then adding or subtracting the scaled temperature probe signal from the signal at the working electrode 58. In this manner, the temperature probe 66 can provide a temperature adjustment for the output from the working electrode 58 to offset the temperature dependence of the working electrode 58.

One embodiment of the temperature probe includes probe leads 68, 70 formed as two spaced-apart channels with a temperature-dependent element 72 formed as a cross-channel connecting the two spaced-apart channels, as illustrated in Figure 8. The two spaced-apart channels contain a conductive material, such as a metal, alloy, semimetal, degenerate semiconductor, or metallic compound. The cross-channel may contain the same material (provided the cross-channel has a smaller cross-section than the two spaced-apart channels) as the probe leads 68, 70. In other embodiments, the material in the cross-channel is different than the material of the probe leads 68, 70.

One exemplary method for forming this particular temperature probe includes forming the two spaced-apart channels and then filling them with the metallic or alloyed conductive material. Next, the cross-channel is formed and then filled with the desired material. The material in the cross-channel overlaps with the conductive material in each of the two spaced-apart channels to form an electrical connection.

For proper operation of the temperature probe 66, the temperature-dependent element 72 of the temperature probe 66 can not be shorted by conductive material formed between the two probe leads 68, 70. In addition, to prevent conduction between the two probe leads 68, 70 by ionic species within the body or sample fluid, a covering may be provided over the temperature-dependent element 72, and preferably over the portion of the probe leads 68, 70 that is implanted in the patient. The covering may be, for example, a non-conducting film disposed over the temperature-dependent element 72 and probe leads 68, 70 to prevent the ionic conduction. Suitable non-conducting films include, for example, Kapton™ polyimide films (DuPont, Wilmington, DE).

Another method for eliminating or reducing conduction by ionic species in the body or sample fluid is to use an ac voltage source connected to the probe leads 68, 70. In this way, the positive and negative ionic species are alternately attracted and repelled during each half cycle of the ac voltage. This results in no net attraction of the ions in the body or sample fluid to the temperature probe 66. The maximum amplitude of the ac current through the temperature-dependent element 72 may then be used to correct the measurements from the working electrodes 58.

The temperature probe can be placed on the same substrate as the electrodes. Alternatively, a temperature probe may be placed on a separate substrate. In addition, the temperature probe may be used by itself or in conjunction with other devices.

## Biocompatible Layer

An optional film layer 75 is formed over at least that portion of the sensor 42 which is subcutaneously inserted into the patient, as shown in Figure 9. This optional film layer 74 may serve one or more functions. The film layer 74 prevents the penetration of large biomolecules into the electrodes. This is accomplished by using a film layer 74 having a pore size that is smaller than the biomolecules that are to be excluded. Such biomolecules may foul the electrodes and/or the sensing layer 64 thereby reducing the effectiveness of the sensor 42 and altering the expected signal amplitude for a given analyte concentration. The fouling of the working electrodes 58 may also decrease the effective life of the sensor 42. The biocompatible layer 74 may also prevent protein adhesion to the sensor 42, formation of blood clots, and other undesirable interactions between the sensor 42 and body.

For example, the sensor may be completely or partially coated on its exterior with a biocompatible coating. A preferred biocompatible coating is a hydrogel which contains at least 20 wt.% fluid when in equilibrium with the analyte-containing fluid. Examples of suitable hydrogels are described in U.S. Patent No. 5, 593, 852, incorporated herein by reference, and include crosslinked polyethylene oxides, such as polyethylene oxide tetraacrylate.

## Interferent-Eliminating Layer

An interferent-eliminating layer (not shown) may be included in the sensor 42. The interferent-eliminating layer may be incorporated in the biocompatible layer 75 or in the mass transport limiting layer 74 (described below) or may be a separate layer. Interferents are molecules or other species that are electroreduced or electrooxidized at the electrode, either directly or via an electron transfer agent, to produce a false signal.

In one embodiment, a film or membrane prevents the penetration of one or more interferents into the region around the working electrodes 58. Preferably, this type of interferent-eliminating layer is much less permeable to one or more of the interferents than to the analyte.

The interferent-eliminating layer may include ionic components, such as Nafion®, incorporated into a polymeric matrix to reduce the permeability of the interferent-eliminating layer to ionic interferents having the same charge as the ionic components. For example, negatively charged compounds or compounds that form negative ions may be incorporated in the interferent-eliminating layer to reduce the permeation of negative species in the body or sample fluid.

Another example of an interferent-eliminating layer includes a catalyst for catalyzing a reaction which removes interferents. One example of such a catalyst is a peroxidase. Hydrogen peroxide reacts with interferents, such as acetaminophen, urate, and ascorbate. The hydrogen peroxide may be added to the analyte-containing fluid or may be generated *in situ*, by, for example, the reaction of glucose or lactate in the presence of glucose oxidase or lactate oxidase, respectively. Examples of interferent eliminating layers include a peroxidase enzyme crosslinked (a) using gluteraldehyde as a crosslinking agent or (b) oxidation of oligosaccharide groups in the peroxidase glycoenzyme with $NaIO_4$, followed by coupling of the aldehydes formed to hydrazide groups in a polyacrylamide matrix to form hydrazones are describe in U.S. Patents Nos. 5,262,305 and 5,356,786, incorporated herein by reference.

**Mass Transport Limiting Layer**

A mass transport limiting layer 74 may be included with the sensor to act as a diffusion-limiting barrier to reduce the rate of mass transport of the analyte, for example, glucose or lactate, into the region around the working electrodes 58. By limiting the diffusion of the analyte, the steady state concentration of the analyte in the proximity of the working electrode 58 (which is proportional to the concentration of the analyte in the body or sample fluid) can be reduced. This extends the upper range of

analyte concentrations that can still be accurately measured and may also expand the range in which the current increases approximately linearly with the level of the analyte.

It is preferred that the permeability of the analyte through the film layer 74 vary little or not at all with temperature, so as to reduce or eliminate the variation of current with temperature. For this reason, it is preferred that in the biologically relevant temperature range from about 25°C to about 45°C, and most importantly from 30°C to 40°C, neither the size of the pores in the film nor its hydration or swelling change excessively. Preferably, the mass transport limiting layer is made using a film that absorbs less than 5 wt.% of fluid over 24 hours. This may reduce or obviate any need for a temperature probe. For implantable sensors, it is preferable that the mass transport limiting layer is made using a film that absorbs less than 5 wt.% of fluid over 24 hours at 37°C.

Particularly useful materials for the film layer 74 are membranes that do not swell in the analyte-containing fluid that the sensor tests. Suitable membranes include 3 to 20,000 nm diameter pores. Membranes having 5 to 500 nm diameter pores with well-defined, uniform pore sizes and high aspect ratios are preferred. In one embodiment, the aspect ratio of the pores is preferably two or greater and more preferably five or greater.

Well-defined and uniform pores can be made by track etching a polymeric membrane using accelerated electrons, ions, or particles emitted by radioactive nuclei. Most preferred are anisotropic, polymeric, track etched membranes that expand less in the direction perpendicular to the pores than in the direction of the pores when heated. Suitable polymeric membranes included polycarbonate membranes from Poretics (Livermore, CA, catalog number 19401, 0.01 μm pore size polycarbonate membrane) and Corning Costar Corp. (Cambridge, MA, Nucleopore™ brand membranes with 0.015 μm pore size). Other polyolefin and polyester films may be used. It is preferred that the permeability of the mass transport limiting membrane changes no more than 4%, preferably, no more than 3%, and, more preferably, no more than 2%, per °C in the

range from 30°C to 40°C when the membranes resides in the subcutaneous interstitial fluid.

In some embodiments of the invention, the mass transport limiting layer 74 may also limit the flow of oxygen into the sensor 42. This can improve the stability of sensors 42 that are used in situations where variation in the partial pressure of oxygen causes non-linearity in sensor response. In these embodiments, the mass transport limiting layer 74 restricts oxygen transport by at least 40%, preferably at least 60%, and more preferably at least 80%, than the membrane restricts transport of the analyte. For a given type of polymer, films having a greater density (e.g., a density closer to that of the crystalline polymer) are preferred. Polyesters, such as polyethylene terephthalate, are typically less permeable to oxygen and are, therefore, preferred over polycarbonate membranes.



**Anticlotting agent**

An implantable sensor may also, optionally, have an anticlotting agent disposed on a portion the substrate which is implanted into a patient. This anticlotting agent may reduce or eliminate the clotting of blood or other body fluid around the sensor, particularly after insertion of the sensor. Blood clots may foul the sensor or irreproducibly reduce the amount of analyte which diffuses into the sensor. Examples of useful anticlotting agents include heparin and tissue plasminogen activator (TPA), as well as other known anticlotting agents.

The anticlotting agent may be applied to at least a portion of that part of the sensor 42 that is to be implanted. The anticlotting agent may be applied, for example, by bath, spraying, brushing, or dipping. The anticlotting agent is allowed to dry on the sensor 42. The anticlotting agent may be immobilized on the surface of the sensor or it may be allowed to diffuse away from the sensor surface. Typically, the quantities of anticlotting agent disposed on the sensor are far below the amounts typically used for treatment of medical conditions involving blood clots and, therefore, have only a limited, localized effect.

**Sensor Lifetime**

The sensor 42 may be designed to be a replaceable component in an *in vivo* or *in vitro* analyte monitor, and particularly in an implantable analyte monitor. Typically, the sensor 42 is capable of operation over a period of days. Preferably, the period of operation is at least one day, more preferably at least three days, and most preferably at least one week. The sensor 42 can then be removed and replaced with a new sensor. The lifetime of the sensor 42 may be reduced by the fouling of the electrodes or by the leaching of the electron transfer agent or catalyst. These limitations on the longevity of the sensor 42 can be overcome by the use of a biocompatible layer 75 or non-leachable electron transfer agent and catalyst, respectively, as described above.

Another primary limitation on the lifetime of the sensor 42 is the temperature stability of the catalyst. Many catalysts are enzymes, which are very sensitive to the ambient temperature and may degrade at temperatures of the patient's body (e.g., approximately 37°C for the human body). Thus, robust enzymes should be used where available. The sensor 42 should be replaced when a sufficient amount of the enzyme has been deactivated to introduce an unacceptable amount of error in the measurements.

The present invention should not be considered limited to the particular examples described above, but rather should be understood to cover all aspects of the invention as fairly set out in the attached claims. Various modifications, equivalent processes, as well as numerous structures to which the present invention may be applicable will be readily apparent to those of skill in the art to which the present invention is directed upon review of the instant specification. The claims are intended to cover such modifications and devices.